Reconsidering the Concept of the Aided Threshold for Nonlinear Hearing Aids

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The aided threshold (and functional gain) has been discussed in the context of linear hearing aids since the early 1960s. The use of nonlinear hearing aids, however, could change the meaningfulness of this verification tool because of their unique characteristics. The interpretation of the aided threshold (and functional gain) as it pertains to linear and nonlinear hearing aids is reviewed. Also discussed are the ideas of an optimal aided threshold, factors that may affect its magnitude, and a comparison between functional gain and insertion gain measures. Finally, how to improve the accuracy of the aided thresholds (and functional gain) through the use of in-situ unaided threshold measurements is discussed.

Introduction

The aided threshold (and functional gain) has been used as a behavioral verification tool since the early 1960s to document the improvement in hearing sensitivity with hearing aid use. Clinicians judge the adequacy of amplification by examining if the aided threshold reaches a prescribed target value. The functional gain provided by the hearing aids, or the difference in audiometer dial readings between the unaided and the aided thresholds, was suggested to reflect gain for sounds presented at a conversational input level (Pascoe, 1975). The measurement of functional gain is needed because the coupler gain of a hearing aid is different from the gain that the wearer receives when it is worn. With the advent of probe microphone real-ear measurement (REM), the value of the behaviorally aided threshold measure is questioned (eg, Stelmachowicz et al., 2002).

There may be many reasons for this change in practice. REM is an "objective" measure capable of reflecting the gain/output at different input

levels on hearing aids that use different signal processing algorithms (Mueller *et al.*, 1992). Furthermore, REM has many applications, such as the measurement of the occlusion effect (Revit, 1992) or directional microphone technology (Ricketts, 2001). Furthermore, the magnitude of the measured insertion gain (for conversational input) was reported to be similar to the magnitude of the measured functional gain (Mason and Popelka, 1986). These examples of the applications of REM give the appearance that REM could provide information that sound-field measures provide and more. In this paper, we explain that each tool has unique applications, and both should be used for hearing aid verification.

The advent of nonlinear signal processing and the application of digital techniques in hearing aids bring new opportunities as well as challenges. One of the challenges is the appropriateness of using behavioral sound-field measurement for verification of hearing aid performance. Today, as many as 80% of audiologists working in an educational setting (Tharpe *et al.*, 2001) and over 50% of dispensing clinicians (Mueller, 1997) routinely perform sound-field measurements for verification. Yet, many discussions on the concept of functional gain and aided thresholds were made in reference to linear hearing aids (Hawkins and Haskell, 1982; Hawkins and Schum, 1984).

The use of nonlinear signal processing can alter the interpretation of the aided threshold (and functional gain) that was discussed in the context of linear hearing aids. Furthermore, the approach to determine this index may need to be modified for nonlinear hearing aids because the measured outcome depends on the properties of the signal used for the measurement (eg, Kuk and Ludvigsen, 2003). Thus, a reexamination of the concept (and the usefulness) of the aided threshold for appropriate application and interpretation is needed. This paper examines the meaning and applications of the aided threshold (and functional gain) as it pertains to linear and nonlinear hearing aids and discusses its relation to insertion gain measures, the factors that may affect its magnitude, and how this index may be predicted.

What Does the Aided Threshold Represent?

Threshold is a behavioral response that reflects the lowest sound pressure level (SPL) at which a listener barely detects the presence of a sound. In a clinical situation, the unaided threshold represents the lowest dial setting that produces a signal at the eardrum that reaches the threshold criterion. In an aided condition, this represents the lowest dial setting that produces an input (I) to the hearing aid microphone which, when added to its in situ gain, results in an output (O) at the wearer's eardrum that reaches the threshold criterion. The difference in dial settings between the unaided and the aided thresholds is defined as the functional gain provided by the hearing aid at a specific frequency. The importance of this definition is that the aided threshold is a behavioral response at one perceptual level (ie, threshold). For a given hearing aid gain setting and test environment, there is only one aided threshold and one value of functional gain. It also means that although the wearer may experience a higher hearing aid output in the ear when the input is higher, the aided threshold (and functional gain) remains the same.

## Expressing the Aided Threshold on a Simplified Input/Output curve

One can use an I/O curve at a particular frequency to illustrate the concept of aided threshold (and functional gain). Typically, an I/O curve reflects the relationship between the input signal in dB SPL (re: 20 μPa/m^2) measured at the hearing aid microphone and its output in dB SPL measured in a 2-cc coupler. This representation does not reflect the actual sound pressure level measured while a person wears the hearing aid. Acoustic transformation would make the sound pressure level measured in the free field or at the microphone position different from that at the eardrum. To circumvent additional acoustic transformation, we will express *the input and the output as the sound pressure level in dB hearing level (HL) at the eardrum,* ie, as the SPL measured at the average wearer's eardrum relative to the average minimum audible pressure (MAP) or the SPL measured in the free field relative to the average minimum audible field (MAF). This is achieved by applying appropriate average corrections/transformations to the input and the output of the hearing aid (see Appendix for a description). The advantage of this representation is that it would have considered any transformation or modification to the input and output by the signal pathway or by the type of transducer used. That

is, the reported sound pressure level on the I/O curve is the "real ear" SPL that would not be modified further. The gain reported from the I/O curve, consequently, will be the real-ear or insertion gain for the average individual.

Expressing the input and the output of a hearing aid in dB HL instead of dB SPL measured at the hearing aid microphone and a 2-cc coupler has the advantage of simplicity. The dB HL scale is identical to the convention in which hearing loss is expressed or measured during routine audiometric practice. It is referenced to how people with normal hearing hear, and its value is independent of the test frequency or the manner in which threshold is determined (ie, in a sound field, under a headphone, or inserts). A person with normal hearing will have a 0-dB HL threshold at all audiometric frequencies regardless of the transducers that are used to determine this threshold value. On the other hand, the person with normal hearing will have different SPL values for different frequencies depending on frequency and how (sound field, earphone) thresholds are measured.

Another advantage of this representation is that it simplifies reporting. For example, if the dB HL notation is used, a person with a 30-dB hearing loss would need an input of 30 dB HL or an output from a hearing aid of 30 dB HL to reach threshold. On the other hand, if the conventional dB SPL scale is used, the same person with a 30-dB HL at 1000 Hz would require an input of 35.7 dB SPL measured in the free field or 38.3 dB SPL measured at the eardrum to reach threshold in an unaided condition. To reach the threshold in an aided condition, an input of 36 dB SPL at the microphone of a transparent behind-the-ear hearing aid or 33.1 dB SPL coupler output from the transparent hearing aid would be needed. A different set of dB SPL values would apply at other frequencies.

Figure 1 shows the I/O curves of a linear hearing aid at various gain settings (20 to 80 dB) for a specific frequency. Through the center (0,0) of the X-,Y-axes is a dashed diagonal line that shows the output in dB HL of the hearing aid measured at the eardrum is the same as its input in dB HL. This shows that neither gain nor attenuation is provided by the presence of the hearing aid, ie, unity gain (or 0-dB gain). Since both the input and the output are referenced to the eardrum, the difference between input and output represents the insertion gain at a particular
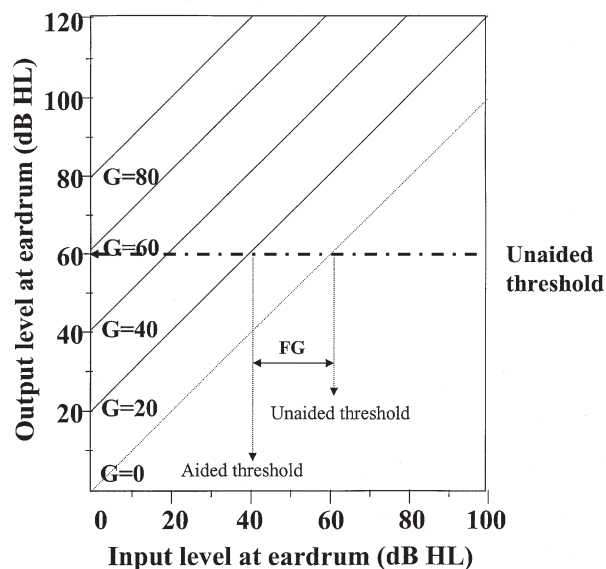
Figure 1. Hypothetical input/output curves of a linear hearing aid illustrating the relationship between unaided threshold, aided threshold, and functional gain (FG). Note that for the same unaided threshold, aided threshold decreases as gain (G) increases.

input level. To the left of the unity gain line are insertion gain lines of increasing magnitude (20 to 80 dB in 20-dB steps). The amount of insertion gain is represented by the magnitude of the Y-axis intercept.

The unaided real-ear threshold of the wearer can be represented by drawing a horizontal line parallel to the X-axis in Figure 1 (dotted-dashed line). This line would represent the sound pressure level in dB HL at the eardrum that signals a threshold response. Consequently, the area above this line would represent audibility, whereas the area below this line would indicate inaudibility. A threshold line at 60 dB HL will intersect the various insertion gain lines at different input levels. Starting from the right-most diagonal line (G = 0 dB or unity insertion gain), these two lines intersect at an input level of 60 dB HL. Because this line is the case of unity gain, the corresponding input level where threshold occurs is the unaided threshold.

One can estimate the aided threshold by determining the input level at which the 60-dB HL threshold line intersects with the insertion gain line. For example, the first line to the left of the unity gain line (ie, G = 20 line) intersects with

the 60-dB HL threshold line at an input level of 40 dB HL. This corresponds to an aided threshold of 40 dB HL. Since there is no lower input level at the same gain setting where threshold occurs, this suggests that *the aided threshold represents the lowest SPL that is required by the wearer to detect the presence of the input signal.* Applying the definition that functional gain is the difference between the unaided and the aided thresholds, this translates to a functional gain of 20 dB (60 dB HL – 40 dB HL), or *an improvement in auditory sensitivity of 20 dB.*

Figure 1 also shows that as the gain of the hearing aid increases (as the solid line shifts to the left), the aided threshold decreases and the functional gain increases. This suggests that the magnitude of the aided threshold is inherently tied to the gain setting of the hearing aid. The hypothetical change in the aided threshold is similar to the change in the volume control setting of the transparent hearing aid (eg, aided threshold decreases from 40 dB HL to 20 dB HL as hearing aid gain increases from 20 dB to 40 dB).

The effect of hearing aid gain on the aided threshold applies to nonlinear wide dynamic range compression hearing aids (WDRC) as well. In contrast to linear hearing aids, where the change in aided threshold (or functional gain) is similar to the change in gain setting, the change in functional gain may exceed the change in the insertion gain setting due to the compressive nature of the I/O curve. As an example, Figure 2 uses a WDRC hearing aid to show that a hearing aid insertion gain of 20 dB (for low input levels) yields an aided threshold at 60 dB HL for a 70-dB hearing loss (or a functional gain of 10 dB). However, increasing the low-level insertion gain to 40 dB yields an aided threshold at 30 dB HL, or a functional gain of 40 dB. In a linear hearing aid, it would have resulted in a 20-dB change in functional gain.

Where Should the Ideal Aided Threshold Be?

Figures 1 and 2 show that for both linear and nonlinear hearing aids, the aided threshold reflects the softest sound that the hearing aid wearer hears (ie, auditory sensitivity of the individual) at a particular gain setting. If the aided threshold were used as an index for hearing aid fitting, it
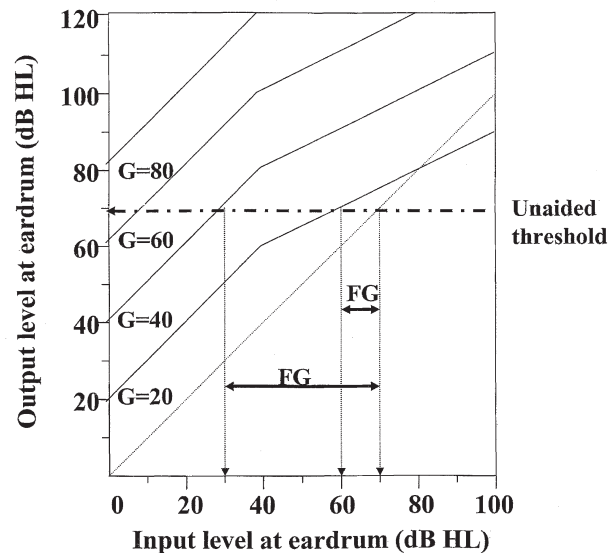
Figure 2. Hypothetical input/output curves of a nonlinear hearing aid illustrating the relationship between unaided threshold, aided threshold, and functional gain (FG). Note that the amount of functional gain varies depending on the gain (G) setting of the hearing aid.

would be logical to ask where the ideal aided threshold should be. A confounding factor in such a consideration is the presence of the volume control. Because changes in the gain setting of the hearing aids affect the aided threshold, one must interpret the results of the aided threshold measurement along with the effect of wearer gain adjustment (ie, volume control). For example, a volume control that allows a gain adjustment of ±10 dB would suggest that the "real world" aided threshold could deviate from the determined one by that magnitude, as long as the output does not exceed the maximum output limit of the hearing aid. To simplify the discussion, the following example assumes no further wearer adjustment on the recommended settings of the hearing aid in real life, ie, a volume control is not present or fixed. This is not unreasonable given that many high-performance digital hearing aids and analog nonlinear hearing aids do not have a volume control. Changes in gain levels on such hearing aids are achieved through level-specific automatic gain control algorithms.

If hearing aids were to enable the hearing impaired person to hear the softest sound comparable to someone with normal hearing, then they

must provide sufficient gain to overcome the threshold displacement that results from a hearing loss. If one can assume that the performance of a hearing aid is not limited by physical constraints, such as circuit noise and acoustic feedback, and that there is no extraneous noise in the environments (ie, we all live in an anechoic chamber), then the theoretical aided threshold should be 0 dB HL because the normal threshold is defined at 0 dB HL. Consequently, 40 dB of functional gain must be provided to someone with a 40-dB hearing loss and 80 dB to someone with a 80-dB hearing loss so that both can hear a sound at 0 dB HL as would a listener with normal hearing. In other words, the theoretical aided threshold should be 0 dB HL for all magnitudes of hearing loss. In order to achieve this, functional gain equivalent to the degree of hearing loss must be provided at an input level of 0 dB HL. Gain of the hearing aid should decrease for input levels above 0 dB HL to compensate for recruitment.

Although theoretically reasonable, having an aided threshold at 0 dB HL may not be feasible or practical. This would require the hearing aid gain to equal the wearer's hearing loss. Clinical experience suggests that mirroring of the audiogram has not been acceptable to wearers of linear hearing aids, who accepted substantially less gain than the mirrored gain. One limiting factor is that sounds at a high input level could become intolerably loud or saturate the hearing aids. Any ambient noise would be amplified by the high gain and become distracting. The stability of the hearing aid may also be an issue because feedback becomes more likely with high gain (Kuk, 1999). These practical considerations would suggest that the ideal, yet practical aided threshold may have to be higher than 0 dB HL, depending on the microphone noise floor and feedback issues of the hearing aid.

Ignoring these technical reservations, one cannot assume that all hearing aid wearers would always prefer a low aided threshold. Dillon *et al.* (1998), using a single-channel fast-acting compression hearing aid, reported that adult listeners preferred substantially less gain at low input levels than might be predicted from a normal aided threshold. It is conceivable, however, that the verbal instructions given to the subjects, the number of channels, and the choice of dynamic parameters of the compression system (eg, attack and release times, expansion versus linear pro-

cessing below the compression threshold) would influence the results significantly. In addition, the perceived importance of needing to hear the low input sounds could affect the outcome. Adults may not prefer to hear all the low input sounds; whereas, children may not develop appropriate speech and language behaviors if they do not hear such low input levels. No consensus has been reached on just how low the practical aided threshold should be.

Alternatively, one may consider the level of meaningful sounds in everyday lives as a practical guide to determining the optimal aided threshold. Assuming that the primary function of hearing aids is to improve speech comprehension, the ideal aided threshold should at least allow the lowest intensity level of normal speech to be audible to the wearer. Although this level may be difficult to define because the measured intensity level is dependent on the duration of the analysis window, the intensity level may be estimated by making certain assumptions. If one assumes that an analysis window of 125 milliseconds is a reasonable duration and make the measurement by using a 1/3-octave bandwidth filter, one would obtain the speech spectrum shown in Figure 3 after it has been transformed from the dB SPL to
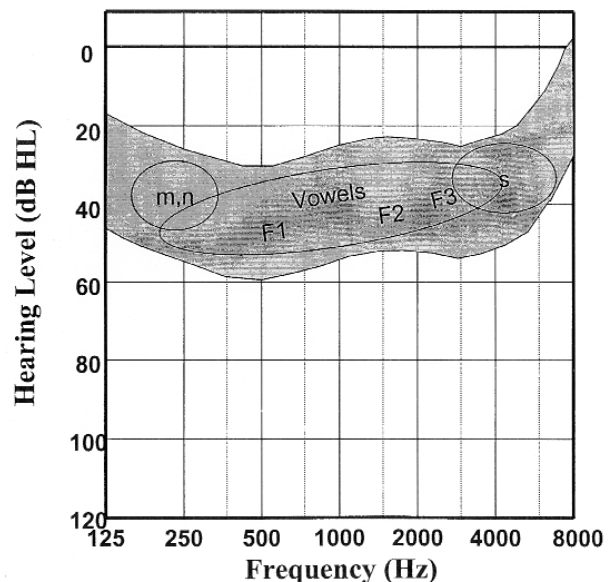
Figure 3. An audiogram with the average speech spectrum superimposed. Note the softer part of average speech occurs at around a 20-dB hearing level (HL) across frequencies.

dB HL scale. In this case, the lowest level of the normal speech spectrum is around 20 dB HL across frequencies. Bearing in mind that the aided threshold reflects the softest sound that the wearer can hear with the hearing aids, Figure 3 suggests that a reasonable aided threshold should be equal to or better than 20 dB HL across frequencies. An aided threshold that is lower than 20 dB HL may ensure audibility for even shorter and softer speech stimuli.

Would the Type of Processing Matter?

The type of processing could affect the acceptability of the aided threshold at 20 dB HL. For a linear hearing aid that provides equal gain at all input levels, achieving a low aided threshold would require high gain from the hearing aid for a wearer who has more than a moderate-to-severe degree of hearing loss. Although the gain for soft sounds may be adequate, the gain for medium and loud sounds may be excessive if no volume control adjustment is allowed. This could cause discomfort and could increase the risk of additional hearing loss (Macrae, 1995; Markle and Zaner, 1966) from overamplification.

Because a person with a mild-to-moderate hearing loss would typically require less gain, achieving an aided threshold at 20 dB HL (or lower) may be appropriate and is possible for mild-to-moderate losses. An aided threshold that allows audibility of average input level sounds (eg, around 40 dB HL) with a linear hearing aid may be more appropriate for people with a more severe hearing loss because of gain limitation on the hearing aid and potential overamplification at high input levels.

Linear hearing aids have an additional problem: Although their gain settings may be optimal for conversational inputs, manual volume control adjustment is necessary for softer and louder sounds. Such an action changes the gain setting of the hearing aid in real life and leads one to question the value of the aided threshold. Consequently, although the aided threshold reflects the softest audible sound to the wearer, frequent gain changes in real-world use of linear hearing aids from volume control adjustment would invalidate the observation that the aided threshold reflects the softest audible sound to the linear hearing aid wearer.

An aided threshold at 20 dB HL or lower may not be as problematic for a nonlinear (or WDRC)

hearing aid as it may be for a linear hearing aid, because a WDRC hearing aid reduces its gain as input increases beyond the compression threshold. At a high input level, the output of a WDRC hearing aid that has the same gain as a linear hearing aid at a low or average input level is likely to be lower than the linear case. This is indicated in Figure 4, which shows the output difference between a linear and a WDRC hearing aid that have the same aided threshold (same gain at a low input level). Thus, a lower aided threshold should be achieved with a WDRC hearing aid over a linear one in order to achieve extra audibility while minimizing the risk of discomfort and potential overamplification. Obviously, an evaluation for potential discomfort is still necessary. For WDRC hearing aids where no manual gain adjustment is possible, the aided threshold does reflect the softest audible sound that the wearer can hear in a quiet environment.

Applications

Assuming that the softest audible meaningful sounds occur at around 20 dB HL, this value can be used as the desired aided threshold for the
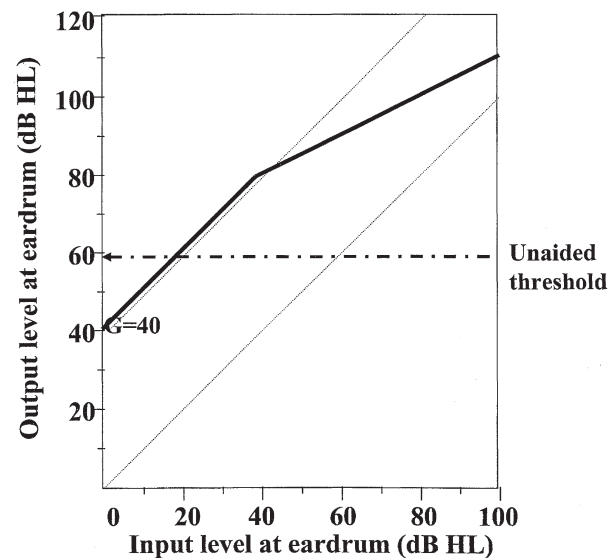
Figure 4. The hypothetical output difference between a linear (dotted line) and a nonlinear (solid line) hearing aid that have the same aided threshold (around 20 dB HL). Note the gain decrease in the nonlinear hearing aid above 40 dB HL.

restoration of audibility. Ignoring the dynamic processing of a nonlinear WDRC hearing aid (which must also be considered), a practical initial goal in a hearing aid fitting is to obtain aided thresholds of around 20 dB HL, regardless of the magnitude and configuration of the hearing loss (with exceptions that follow). The measured-aided threshold can be compared with this 20-dB HL target to examine the "goodness" of the fitting of nonlinear hearing aids. The value of the aided threshold can be used as a guide for setting the gain for low-level inputs to the hearing aid. Once obtained, this information (of the aided threshold) can be used along with the speech spectrum diagram (Figure 3) to explain to the hearing aid wearers the lowest input sounds they can hear.

Practically, because of gain limitation and feedback issues, it may be possible only for people with up to a moderately severe hearing loss to achieve this normal aided threshold. One may have to accept a higher level (eg, 30 to 40 dB HL) for people with a severe-to-profound hearing loss unless effective means to prevent feedback is available. Also, recent reports on desensitization and dead regions (eg, Ching *et al.*, 1998; Hogan and Turner, 1998; Moore, 2001) would suggest that at least for adults, it may not be desirable to achieve a low aided threshold when the hearing loss exceeds 60 to 80 dB HL and for certain hearing loss configurations (eg, steeply sloping or rising losses). In addition, individuals with severe hyperacusis may not accept the desired gain. Aside from these caveats, it seems reasonable as a starting point to aim for a low aided threshold with nonlinear hearing aids in order to maximize audibility. Obviously, empirical evidence is needed to support this assertion and to define the exceptions to this rule, especially with its applications on multichannel hearing aids (eg, Kuk and Ludvigsen, 2003).

Although the value of the aided threshold as a fundamental measure of hearing sensitivity for nonlinear amplification may be indisputable, one must guard against overly extending its interpretation. Because the aided threshold represents the softest sounds that are perceptible, the calculated functional gain for a nonlinear hearing aid would not reflect gain for conversational sounds as would be suggested for linear hearing aids. Furthermore, the value of the aided threshold (and functional gain) does not reflect or predict speech intelligibility skills (either in quiet or noise). Neither will the aided threshold indicate if the output of the hearing aids is too high. In addition, the aided threshold does not reflect processing of the hearing aids, such as noise reduction and directional microphones, other than its processing for soft sounds in quiet. Nonetheless, the information provided by this index of audibility is critical because every aspect of auditory processing—from a simple detection task to a complex speech-understanding task—originates from the ability to hear and detect the sound. Without audibility, there can be no intelligibility. Other verification tools, such as probe microphone measures, do not provide such information. The aided threshold is the only verification index that provides such information. Consequently, the aided threshold must be measured if the goal of verification is to assess the softest sound that the wearer hears with the nonlinear hearing aids

Variables Affecting the Aided Threshold (or Functional Gain)

Shape of the Input/Output Curves

Figure 1 illustrated that the aided threshold is the lowest input level where the threshold line intersects with the I/O curve. Consequently, any I/O curves that have the necessary gain at the desired input level of the aided threshold should yield the same aided threshold. The characteristics of the I/O curve above and below that input level should not affect the value of the aided threshold. That does not mean, however, that the performance of the hearing aid is identical in other aspects. For example, Figure 5 shows four possible I/O curves for a hearing loss of 60 dB HL (with a hypothetical loudness discomfort level of 118 dB HL) that has an aided threshold at 20 dB HL.

As in the previous figures, the dashed line in Figure 5 that runs through the origin of the I/O curve is the line of unity gain. Line A (solid) is shifted 40 dB to the left of the line of unity gain. This represents linear processing from 0 dB HL to 70 dB HL with a gain of 40 dB. While an aided threshold of 20 dB HL is achieved, the output indicated by this line is higher than the other three I/O curves above input levels of 20 dB HL (for C and D) and 40 dB HL (for B). This suggests higher output with linear processing than with non-linear processing. In addition, LDL is reached at a
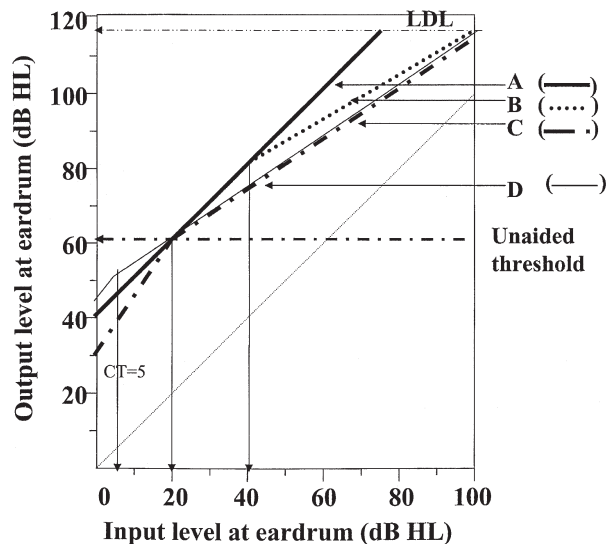
Figure 5. Hypothetical input/output (I/O) curves showing that different I/O characteristics can result in the same aided threshold (at 20 dB HL). Line A shows a linear hearing aid with a 40-dB gain. Line B shows a wide dynamic range compression (WDRC) hearing aid with a compression threshold at 40 dB HL and a low-level gain of 40 dB. Line C shows a WDRC hearing aid with a compression threshold at 20 dB HL and low-level gain of 40 dB. Line D shows a WDRC hearing aid with a compression threshold at 5 dB HL with a 45-dB low-level gain. Abbreviation: CT, compression threshold.

moderate input level (ie, around 70 dB HL). This reinforces the earlier discussion that a low aided threshold may not always be desirable in a linear hearing aid.

Curve B (dotted line) represents a WDRC hearing aid with a low-level gain of 40 dB at a compression threshold of 40 dB HL. Reduced gain is seen for input levels above 40 dB HL. Because the compression threshold is the lowest input level at which gain reduction occurs (Kuk, 2000), the compression threshold also represents the input level with the maximum gain on the input/output curve. Because linear processing is used below the compression threshold, gain at the compression threshold is the same as that below. This suggests that gain at the 20 dB HL input is also 40 dB, sufficient to yield an aided threshold at 20 dB HL. If expansion instead of linear amplification is used below the compression threshold, gain at the 20 dB HL input would be

less than 40 dB and the aided threshold will be higher than 20 dB HL.

Curve C (dash-dot) represents another input compression hearing aid that has the compression threshold at 20 dB HL with a maximum gain of 40 dB. Expansion is used below 20 dB HL to minimize microphone noise. This I/O design targets the compression threshold as the desired aided threshold. Because curve C has a lower compression threshold than curve B, less compression is needed (ie, lower compression ratio) for curve C (than curve B) to retain the range of input within the subject's dynamic range (below LDL).

Curve D (thin solid) illustrates another variation of curve C. Instead of a compression threshold at 20 dB HL, curve D has a compression threshold at 5 dB HL. In order that the input level of 20 dB HL has 40 dB of gain, gain at the compression threshold has to be higher than 40 dB. Despite a higher maximum gain of 45 dB, the output below 20 dB HL is inaudible to the wearer. Above the 20 dB HL input, both curves C and D have the same I/O characteristics (ie, same compression ratio).

A comparison of these four curves shows that although gain below 40 dB HL is identical between curves A and B, curve B has significantly less output above 40 dB HL than curve A, but more output than curves C and D. This allows curve B to reach LDL at a higher input level than curve A (ie, more headroom). However, in order for it to reach the same maximum output level as curves C and D, a higher compression ratio (ie, more gain reduction) is needed in curve B than in curves C and D.

These four curves illustrate that the same normal aided threshold (at 20 dB HL) can be achieved with different types of I/O processing characteristics—from linear processing to nonlinear processing—with different compression thresholds and ratios. Although the aided threshold is the same, the perception of loudness and sound quality resulting from each type of processing could be very different.

Hearing Aid Gain Parameters

Because the effect of gain adjustment on linear hearing aids is observed at all input levels, it is expected that the value of the aided threshold would be affected whenever the gain is adjusted in linear hearing aids. On the other hand, a non-linear hearing aid allows gain adjustment at more

than one input range. Thus, the effect on the aided threshold of adjusting different gain parameters may be different. Because of the variation in processing schemes used in different compression hearing aids, the following discussion may not be applicable to all hearing aids. Readers are urged to refer to the I/O curves of the specific hearing aids to examine how changes in the specific electroacoustic parameter may affect the aided threshold (or functional gain).

Gain for soft sounds—It was indicated earlier that the value of the aided threshold is determined by the amount of gain on the hearing aid at a low input level. Consequently, any changes in the gain for soft sounds could potentially affect the value of the aided threshold. For example, Figure 6a shows that as the compression threshold of a hearing aid is increased from 20 dB HL to 40 dB HL, gain for soft sounds decreases and the aided threshold increases from 20 dB HL to 30 dB HL.
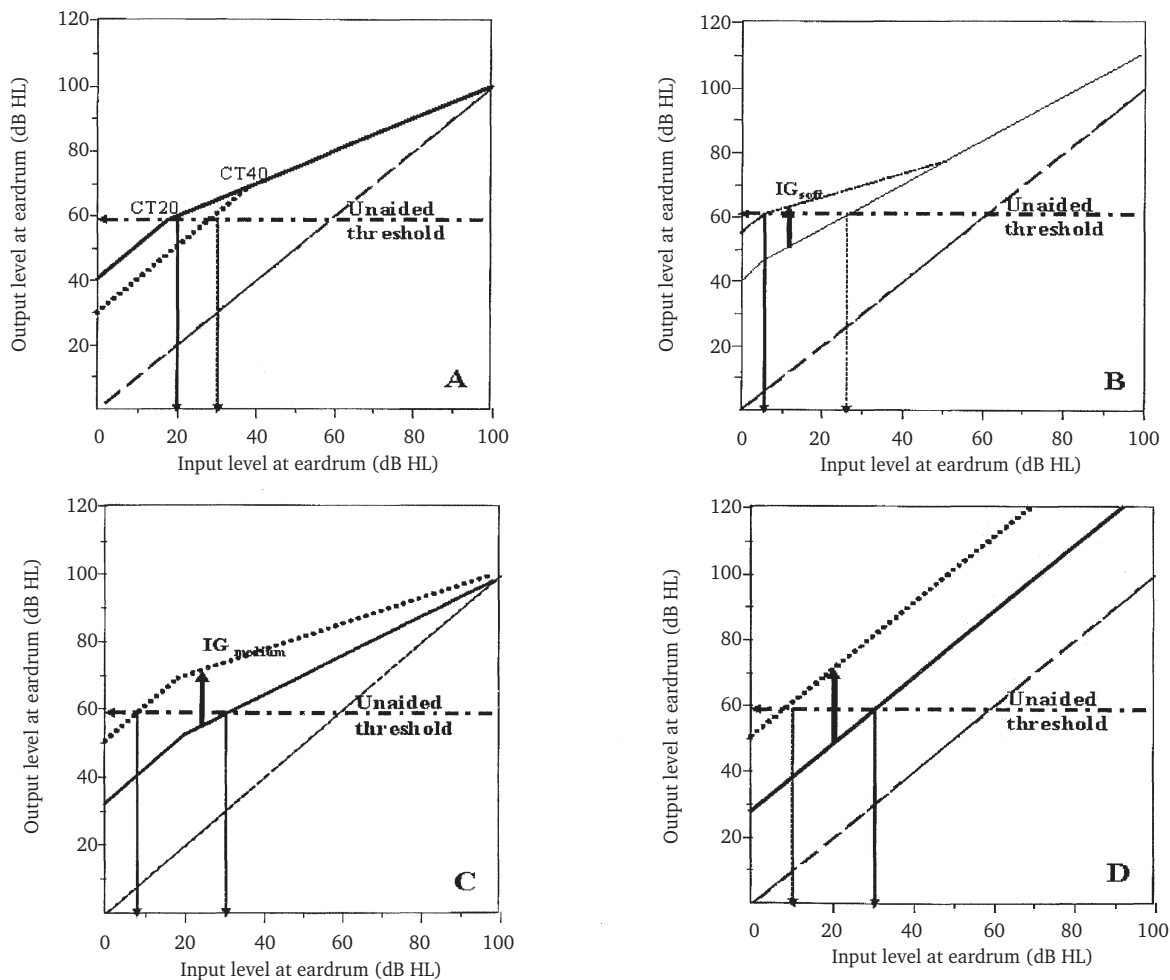
Figure 6. A) Effect of compression threshold on the aided threshold. As the compression threshold increases, gain below the compression threshold decreases and aided threshold increases. B) Effect of soft insertion gain (IG$_{soft}$) on the aided threshold. As IG$_{soft}$ increases, aided threshold decreases. C) Effect of IG medium (from 30 dB to 50 dB) on the hypothetical input/output curve. As IG medium increases, gain increases and aided threshold decreases. D) Effect of IG increase on the aided threshold in a linear hearing aid. As gain increases, aided threshold decreases. Abbreviation: CT, compression threshold.

In another hearing aid (6b), as the gain for soft sounds is increased from 40 dB to 60 dB, gain below a conversational level increases and consequently, the aided threshold decreases from 25 dB HL to 5 dB HL. In yet another case (6c), as the gain parameter for average level or medium sounds (ie, IG$_{medium}$) increases from 30 dB to 50 dB, gain for soft sounds is also increased and the aided threshold is lowered from 30 dB HL to 10 dB HL. As the gain for a linear hearing aid (6d) is increased from 30 dB to 50 dB, its gain for all input levels is increased (including soft sounds) and a decrease in aided threshold from 30 dB HL to 10 dB HL is seen. The impact of some of these changes may not be felt in "typical" real life environments where the ambient noise level (typically around 15 to 20 dB HL measured in the 1/3-octave band) may mask the effect of any parametric changes; however, the changes will be perceptible in environments with low ambient noise level.

Gain for medium or average level sounds—For many nonlinear hearing aids, the gain parameter responsible for medium or conversational sounds includes gain adjustment for lower input sounds as well. In such cases, increasing the gain for average-level sounds would lower the aided threshold. This was seen in Figure 6c. On the other hand, some digital hearing aids allow specific gain adjustment such that gain adjustment for conversational sounds only affects inputs in the range of 35 to 65 dB HL (Figure 7). In such a case, adjustment of this gain parameter would not affect the aided threshold. It is important to consider the effect of adjusting a specific gain parameter in order to realize how it influences gain at different input levels, especially the aided threshold.

Gain for loud sounds—Some nonlinear hearing aids allow independent gain adjustment for loud sounds. Figure 8 shows the effect of adjusting that parameter is mainly at high input levels. No, or minimal, effect is seen at low input levels. Because the aided threshold reflects gain at low input levels, adjusting the IG$_{loud}$ parameter would have no effect on the aided threshold in those hearing aids.

Measurement Variables

Certain measurement variables could affect the value and reliability of the aided threshold (and
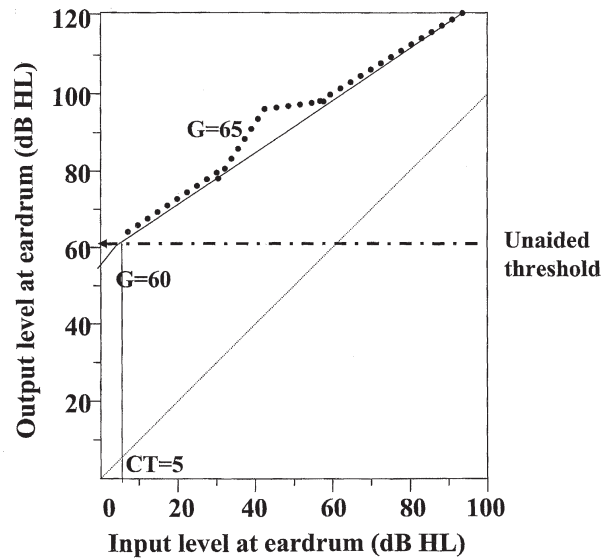
Figure 7. Effect of medium insertion gain (30 to 50 dB) change on a hypothetical input/output curve of a digital hearing aid. Note that gain (G) for input levels below 40 dB HL is not affected. Abbreviation: CT, compression threshold.
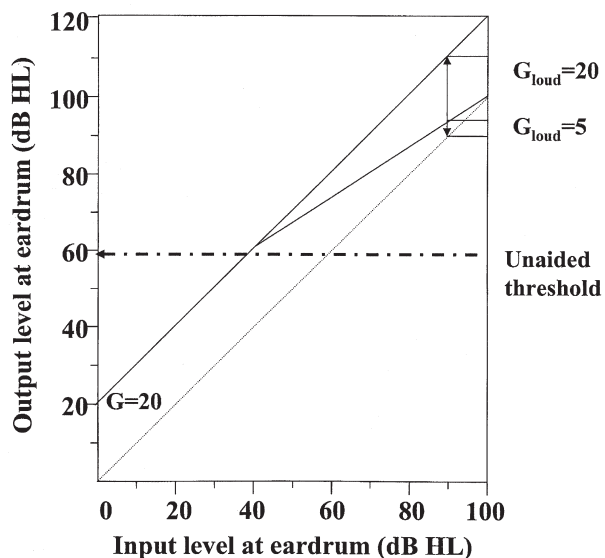
Figure 8. Effect of loud insertion gain (IG) (from 20 dB to 5 dB) on the hypothetical input/output curve. As IG loud increases, gain increases above the compression threshold and aided threshold remains unchanged.

functional gain). Walker (1995) provided an excellent tutorial on the factors that may affect sound-field threshold measurement. Some of these variables include:

1. *Ambient and circuit noise*—Ambient room noise and low-level noise in the test environment may act as maskers and raise the level of both aided and unaided thresholds and alter the magnitude of the functional gain. This would be especially true for frequencies below 500 Hz. Subjects with normal hearing or a mild hearing loss in the low frequency would especially be affected. Consequently, test rooms for conducting sound-field testing must be free from any extraneous noise sources. Furthermore, Macrae and Frazier (1980) also pointed out that circuit noise from hearing aids could also impose a floor effect on the aided threshold and raise its level, and thus decrease functional gain.

2. *Standing waves*—Because most test environments are enclosed, standing waves will likely develop from reflections off of the test booth walls. To overcome this, the use of frequency-modulated pure tones (or warble tones) as the test stimuli is recommended because they cover a narrow frequency region and are less susceptible to room resonance (Walker, 1995).

3. *Head and body movement*—Any movement by the test subject during sound-field testing risks changing the acoustic input to the ear and can result in a threshold shift. The effect may be more significant in the higher frequencies because they have shorter wavelengths.

4. *Volume control adjustment*—A source of variability in aided threshold measurement is that subjects were sometimes allowed to adjust the volume control on the hearing aid (Hawkins *et al.*, 1987). An explanation for this observation was made previously. Subjects should not be allowed to make any volume control adjustment on the hearing aids during testing and upon retesting.

5. *Identical test conditions*—Threshold measurements to determine functional gain (difference between aided and unaided thresholds) must be obtained under identical test conditions. It is inappropriate to compute functional gain by comparing the monaural unaided thresholds obtained under headphones to the aided sound-field thresholds obtained in a binaural manner. The true functional gain provided by a hearing aid is the difference between the monaural unaided threshold and the monaural aided threshold obtained in the same sound-field test environment.

6. *Artifacts with sloping loss*—In a precipitous hearing loss where there is normal hearing in the low frequency and a severe-to-profound loss in the high frequency, intermodulation distortion could occur at the loudspeaker or at the hearing aid at a high output level. This results in an audible distortion product that is much lower in frequency and level. For example, a person with normal hearing up to 500 Hz but with a 80-dB HL at 4000 Hz may indicate a threshold response to a 60-dB HL, 4000-Hz warble tone. This is because as little as a 1% distortion of the 4000-Hz tone will create intermodulation distortion products that are approximately 20 dB at low frequencies (eg, 500 Hz). This distortion product is easily detected by the low-frequency fibers, leading to a false threshold response. Ipsilateral masking may be necessary to prevent this possibility.

7. *Nonlinear hearing aids*—Because of the dynamic change in gain characteristics over time, the timing characteristics (attack and release times) of a nonlinear hearing aid could interact with the stimuli and affect the measured aided thresholds. This may be especially true in slow-acting WDRC hearing aids and in hearing aids with a low compression threshold. Because a compression hearing aid decreases its gain as input level increases beyond the compression threshold, the typical bracketing approach used in threshold estimation may yield more variable results when applied to a nonlinear hearing aid than to a linear hearing aid. Slow-acting compression and WDRC hearing aids with a low-to-moderately low compression threshold are especially susceptible.

A bracketing approach involves a frequent, large-intensity difference (and thus gain difference) between stimulus presentations (eg, up 10 dB and down 5 dB or 15 dB in each "bracket"). Stimuli that are below the compression threshold would not activate the compression circuit to introduce

the output uncertainty. If the stimulus that is used to estimate the aided threshold is at or above the compression threshold of the hearing aid, it will activate compression. It may not be easy to predict the lower output resulting from compression because of the interaction between the time constants of the hearing aids and the stimulus characteristics (see next section). One approach is to present stimuli in an ascending manner in 5-dB steps once the vicinity of the aided threshold is bracketed. This would minimize unpredictable gain swing and its associated variability in the aided threshold measure.

If the stimulus level is above the compression threshold, the duration of the signal could interact with the attack time of the nonlinear hearing aid and affect the aided threshold. With a short attack time, only the initial part of the stimulus may receive the full gain from the hearing aid while the remaining part of the stimulus receives the reduced gain. The drawing on the left side of Figure 9 illustrates the situation. On the other hand, the same stimulus would receive maximum gain from the hearing aid for a longer period of time if it uses a long attack time (right side of Figure 9). A higher output over the duration of the stimulus is possible. A lower aided threshold may result for the same audiometric threshold.
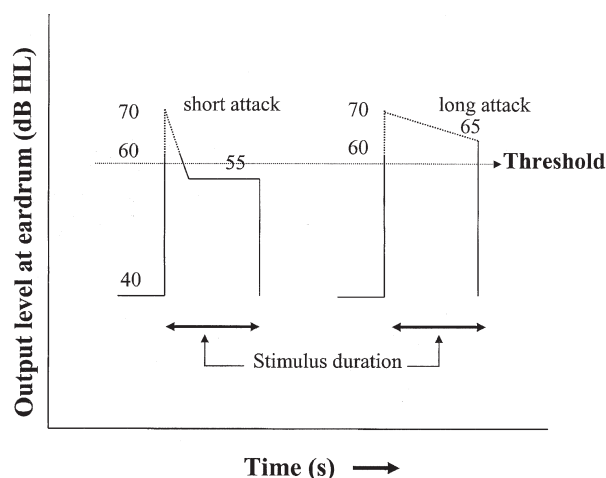
Figure 9. Schematic diagram of the effect of attack time on the output waveform. For the same audiometric threshold, a nonlinear hearing aid with a long attack time (right) could yield a lower aided threshold than one with a shorter attack time (left).

Thus, two hearing aids with identical I/O characteristics could yield different aided thresholds if they are significantly different in their attack times. The hearing aid with a longer attack time would likely yield a lower aided threshold. Given that most WDRC hearing aids use a relatively short attack time (less than 10 milliseconds), it is foreseeable that the aided threshold is measured in a reduced-gain state. For WDRC systems that use a low compression threshold, a long attack time may "slow" gain reduction and yield a better (or lower) aided threshold. To obtain a consistent aided threshold, a brief stimulus (about 1 to 2 seconds) that has a smooth onset might be preferable.

The interval between stimulus presentations could also affect the reliability of the aided threshold in a slow-acting WDRC hearing aid that uses a low compression threshold. Again, for stimuli presented below the compression threshold, no special precautions are necessary because the hearing aid operates on a linear portion of the I/O curve (not true for those with low-level expansion). Consecutive stimuli presented above the compression threshold will typically be presented at different stages of the gain recovery phase of the hearing aid. This means that each stimulus could potentially receive different gain (and output) to yield variable aided thresholds. If possible, one should wait for the duration of the release time before the next presentation to minimize variability from presentations at various gain levels. However, if one uses an ascending approach and assumes that the intensity difference between presentations is typically 5 dB, the minimum required interval between presentations can be as brief as 1 to 2 seconds, even for a slow-acting WDRC hearing aid with a low compression threshold.

In summary, the attack time and the release time on a nonlinear hearing aid may affect the reliability and validity of the obtained aided threshold. A nonlinear hearing aid with a high compression threshold and linear processing below the compression threshold may yield less variability in the aided threshold than one with a lower compression threshold. The use of expansion below the compression threshold may further add variability. Typically, an ascending approach with a stimulus duration of about 1 to 2 seconds and an interval between presentations of 1 to 2 seconds should ensure a reliable outcome.

Predicting the Aided Sound-Field Threshold (and Functional Gain)

The aided threshold is useful as a measure of audibility in nonlinear hearing aids, but endogenous and exogenous factors could affect its reliability. Several authors (Hawkins *et al.*, 1987; Humes and Kirn, 1990; Stuart *et al.*, 1990) reported significant variability on aided thresholds (and functional gain) obtained with linear hearing aids. Indeed, these studies suggest test–retest standard deviations of the aided threshold (and functional gain) ranging from 3 dB to more than 10 dB with a typical deviation of 6 dB. Such variability may increase for nonlinear hearing aids for the reasons cited in the previous section. This variability, when coupled to the limited cooperation from infants and very young children, is another reason why some authors questioned its clinical utility and recommended probe microphone measurements instead (eg, Stelmachowicz *et al.*, 2002). To ensure that the value of the aided threshold is meaningful, it may be desirable to predict its magnitude without patient involvement. This could minimize variability and save valuable time for other productive clinical activities. We will examine this possibility by reviewing what measurements may be needed for this prediction. The following derivation may help in this regard.

Consider a situation in which the test person is placed in a plane, freely progressive sound-field and is facing the sound source. The intensity of the sound source is adjusted until a threshold response is obtained with and without a hearing aid. At threshold, the sound pressure level measured at the eardrum (SPL[ED]) without a hearing aid is related to the source signal generated in the free sound-field (SPL[FF]$_{unaided}$) and the free-field-to-eardrum transform (FF2ED) by Equation 1.

$$SPL(ED)_{unaided} = SPL\ (FF)_{unaided} + FF2ED \quad (1)$$

When a hearing aid is worn, the sound pressure level measured at the eardrum at threshold (SPL[ED]$_{aided}$) is related to the source signal (SPL[FF]$_{aided}$) by the free-field-to-microphone transfer for the specific style of hearing aid (FF2Mic$_{HA\ type}$, also known as microphone location effect (MLE) worn and the in situ gain of the hearing aid. The last variable can be measured directly through real-ear techniques or calculated through knowledge of the coupler gain of the

hearing aid and the individual real-ear-to-coupler difference (RECD) of the wearer.

$$SPL(ED)_{aided} = SPL(FF)_{aided} + FF2Mic_{HA\ type} + coupler\ gain + RECD \quad (2)$$

Assuming that the SPL measured at the eardrum at threshold is the same for both the aided and the unaided conditions, Equation 1 should be the same as Equation 2, giving:

$$SPL(FF)_{unaided} + FF2ED = SPL(FF)_{aided} + FF2Mic_{HA\ type} + coupler\ gain + RECD$$
or
$$SPL\ (FF)_{aided} = SPL(FF)_{unaided} + FF2ED - FF2Mic_{HA\ type} - coupler\ gain - RECD \quad (3)$$

Equation 3 shows that one may be able to predict the sound-field aided threshold of the wearer if one knows the sound-field unaided threshold of the wearer, the free-field-to-eardrum transfer function of the particular wearer, free-field-to-microphone transfer characteristics of the particular style of hearing aid as it is worn on the wearer, the coupler gain of the hearing aid, and the real-ear-to-coupler difference of the individual to account for such things as the effects of venting, volume, and impedance difference, and style of hearing aid.

While these indices can be measured, one cannot but question the real advantage of prediction if one were to measure the unaided sound-field threshold, the FF2Mic, RECD, and FF2ED transforms. The variability that one encounters in real-ear probe microphone measures when determining the RECD or in acoustic measures like FF2Mic and FF2ED may be just as great, if not greater, than the variability seen in aided threshold or functional gain measures (Arlinger and Jerlvall, 1987; Cuda *et al.*, 1992, Valente *et al.*, 1991). Furthermore, the time involved in making these measurements may be prohibitive for clinical purposes. One practical solution for predicting the aided threshold is to use appropriate group correction factors for the specific style of hearing aids (eg, FF2ED, FF2Mic, RECD). For example, Bentler and Pavlovic (1989) provided a summary of typical transforms that are reported in the hearing aid literature. The use of group transforms and corrections simplifies the prediction and provides a reasonable estimate for the average person. One must recognize that individual deviations will occur in some situations to re-

sult in an inaccurate prediction. To increase the accuracy of prediction, one needs to increase the accuracy of the measured unaided threshold and minimize the need for unnecessary transforms.

The use of digital signal processing in some hearing aids may allow greater accuracy in the prediction of the aided thresholds by increasing the accuracy of the measured unaided threshold. These hearing aids are capable of generating acoustic signals that can be used as stimuli for in situ threshold determination (Ludvigsen and Tøpholm, 1997). As part of the fitting process, the hearing-impaired person wears the actual hearing aids (or earmolds coupled to the hearing aids), listens to stimuli that are generated from within the hearing aids (ie, bypassing the microphone), and is instructed to respond in the same manner to the test stimuli as in a routine audiometric threshold determination. The level of the stimuli is calibrated in reference to an ear simulator (IEC 711) with electroacoustic properties that approximate the average adult ear characteristics.

Although the reference is not made to the individual ear, the magnitude of the in situ threshold determined in this manner reflects the combined effects of all the hearing aid acoustic variables (eg, venting, impedance difference, residual volume) that determine the unaided threshold. In essence, this method enables the hearing aid to have its sound reproduction calibrated relative to the individual threshold since the wearer's hearing aid is the actual transducer used for measuring the threshold. Once the threshold is determined, one can specify output levels relative to the threshold (and I/O curve characteristics) on the hearing aid by using an appropriate target/prescription with suitable allowance made for the properties of the hearing aid, such as the number of channels, time constants, and hearing aid styles (Kuk and Ludvigsen, 1999).

This approach allows the aided threshold to be determined as the lowest input level when the output of the hearing aid (input level plus gain) reaches the in situ threshold of the wearer. The process can be simplified even more if one adopts the dB HL (re: eardrum) notation in expressing the input/output relationship (see the Appendix for details).

Figure 1 can be used as an example of how in situ threshold measures can be used to estimate the aided threshold. Assume that the I/O curve for a gain setting of 40 dB is chosen on the hearing aid for the specific frequency of interest.
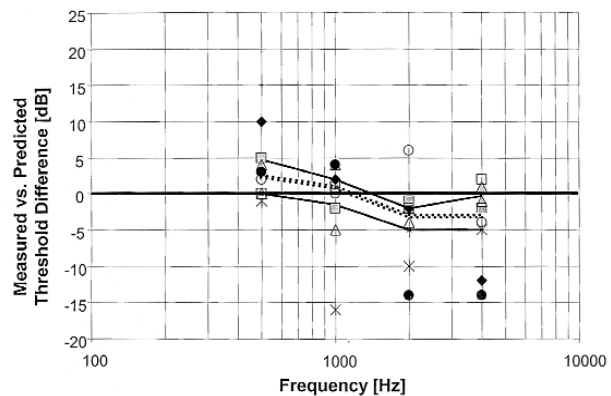
Figure 10. Individual measured versus predicted aided threshold difference across frequencies. The dotted line around 0 is the mean deviation across frequencies. The two black lines are the 25th and 75th percentiles. Each symbol represents a different subject.

If the in situ threshold of the wearer is 60 dB HL, one can draw a horizontal line across the Y-axis where the in-situ output is 60 dB HL. The input level (in dB HL) where the I/O curve intersects with the in situ threshold is the aided threshold. In this case, the predicted aided threshold is 20 dB HL.

A note of caution: Because the unaided in situ threshold is obtained with signals originating from within the hearing aid, the predicted aided threshold should not be taken as the aided sound-field threshold unless the individual free-field-to-microphone (FF2Mic) correction is added to the prediction to account for the effect of the sound transmission pathway. Group FF2Mic transforms for different hearing aid styles will need to be included in the prescribed I/O curves to provide an unbiased estimate of the aided threshold. Although the use of an average FF2Mic correction would not account for individual variations in FF2Mic values, its use is warranted since the range of individual variation in a FF2Mic correction is less than 10 dB across frequencies in a completely-in-the-canal hearing aid (Cornelisse and Seewald, 1997).

This method of measuring a wearer's in situ threshold and predicting the sound-field aided threshold is implemented in the Widex Senso digital hearing aids. As part of the validation, a preliminary study compared the monaurally measured sound field-aided thresholds to the predict-

ed monaural sound-field aided thresholds from the I/O curve in Compass (the fitting software) in 10 hearing-impaired subjects (Kuk *et al.*, 2003). Figure 10 shows the individual deviation between the measured and the predicted aided thresholds at 500, 1000, 2000, and 4000 Hz. Each symbol represents a different subject. Data points that are above 0 suggest that the measured aided thresholds are higher than the predicted aided thresholds, whereas data points below the 0 line suggest that measured aided thresholds are lower than predicted aided thresholds. The dotted line around 0 is the mean deviation across frequencies. The two solid lines represent the 25th and 75th percentile. Most of the subjects showed less than 5 dB deviations between the measured and the predicted aided thresholds. This is especially true for frequencies at 1000 and 2000 Hz. At 500 Hz, the mean measured aided threshold is about 3 to 4 dB higher than the predicted threshold. At 4000 Hz, the average measured aided threshold is about 3 to 4 dB lower than the predicted aided threshold. On the other hand, two subjects (*,•) showed as much as a 15-dB deviation at some frequencies. These observations suggest that the sound-field aided threshold can be predicted with fair accuracy (±5 dB) from the I/O curve. This magnitude of variability is acceptable considering that the variability in sound-field measurement is similar in magnitude. For example, Humes and Kirn (1990) showed a standard deviation of 3 to 4 dB for unaided threshold measures. Furthermore, considering that prediction eliminates the need to measure the sound-field aided thresholds, which may be difficult for uncooperative patients, this level of potential variability is tolerable.
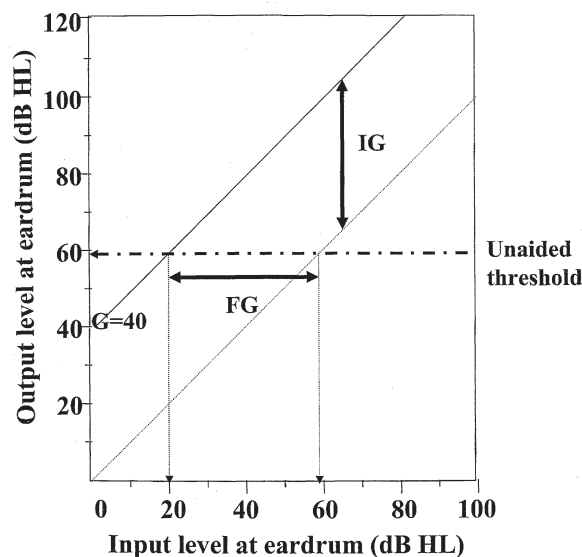
Relationship Between Functional Gain and Insertion Gain

Aided threshold represents the audibility of the softest sounds and has the same meaning for linear and nonlinear hearing aids. The interpretation of functional gain is different between linear and nonlinear hearing aids, especially when its relationship to insertion gain (IG) is considered. By definition, the real-ear insertion gain is the difference in output measured at the eardrum between a real-ear unaided response and a real-ear aided response condition at a specific input level.

For linear hearing aids, Mason and Popelka (1986) are among those who reported that functional gain was similar in value to insertion gain.

Figure 11 illustrates the theoretical relationship between functional gain and insertion gain in a linear hearing aid. Because functional gain is the difference in input levels between the aided and the unaided thresholds, the horizontal distance between the two diagonal lines on the X-axis, one for a gain of 0 and the other for a gain of 40, would represent the value of the functional gain. On the other hand, if insertion gain is the difference in hearing aid output between the aided and the unaided conditions at a particular input level, the vertical distance between the two diagonal lines at a specified input should represent insertion gain. Figure 11 shows that in a linear hearing aid, functional gain should be identical to insertion gain. This hypothetical equivalence of functional gain and insertion gain is valid at different gain settings and for various degrees of hearing loss in a linear hearing aid.

Figure 12 shows a conventional WDRC hearing aid with an insertion gain of 40 dB at and below the compression threshold at 40 dB HL. For a 60-dB hearing loss, the predicted aided threshold is 20 dB HL. This corresponds to a functional gain, barring any measurement errors, of 40 dB.

Figure 11. The hypothetical relationship between functional gain and insertion gain on a linear hearing aid. Note that the two are similar in magnitude.
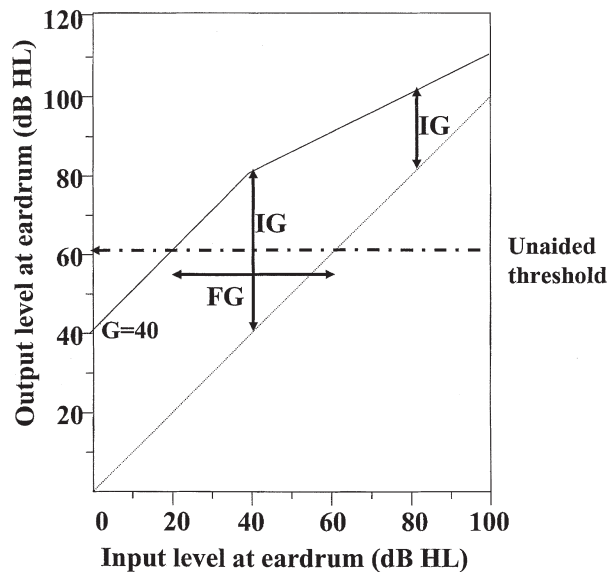
Figure 12. The hypothetical difference between functional gain (FG) and insertion gain (IG) at a gain setting of 40 dB on a nonlinear hearing aid. The two terms are identical for inputs below a compression threshold of 40 dB HL. The two are different for inputs above the compression threshold. Abbreviation: NL, nonlinear.

The insertion gain, however, differs from 40 dB, depending on the level of the input signal. An input signal of 40 dB HL would yield an aided output at 80 dB HL, or an insertion gain of 40 dB. Indeed, the insertion gain will be 40 dB as long as the input level is lower than the compression threshold (in this case, less than 40 dB HL) and assuming that linear processing is used below the compression threshold. At a higher input level (eg, 60 dB HL) the aided output will be 90 dB HL. This is equivalent to an insertion gain of 30 dB. Indeed, as input increases above the compression threshold, insertion gain decreases. Although not shown in Figure 12, one can also appreciate a greater decrease in insertion gain with input level increases for a higher compression ratio. In short, the difference between functional gain and insertion gain increases beyond the compression threshold and for higher compression ratios.

Figure 12 also shows that in cases where the input level is insufficient to elicit a threshold response (eg, 10 dB HL for the insertion gain setting of G = 40 dB), there will not be any aided threshold or functional gain. Yet, the insertion

gain of the hearing aid will be 40 dB. This again stresses the perceptual all-or-none nature of the aided threshold.

Previous research has reported discrepancies between measured insertion gain and functional gain also. Schwartz and Larson (1977), and Seewald *et al.* (1992) reported that the sensation level of amplified speech estimated by the use of functional gain measures was typically greater than insertion gain measures that used a 70 dB composite signal. Although detailed information about the subjects' hearing aids were not provided, based on the degree of hearing loss and the time of the study, it is likely that these hearing aids were linear hearing aids. Such a discrepancy would suggest that the input level used for the insertion gain measure might have exceeded the linear range of the hearing aids (ie, either in compression or saturation) to result in such findings. Stelmachowicz and Lewis (1988) also reported three situations (high gain but low OSPL90, nonlinear hearing aids, and profound hearing loss) in which functional gain was different from insertion gain. Furthermore, sound-field behavioral threshold measurements may underestimate functional gain if unaided thresholds approach the normal hearing range (Rines *et al.*, 1984). This is likely caused by masking from either circuit noise in the hearing aid or background noise present in the test room.

These observations suggest that functional gain and insertion gain should not be interpreted the same way, especially for nonlinear hearing aids. To fit a linear hearing aid optimally, most fitting approaches prescribe gain to amplify conversational speech to the wearer's most comfortable listening level (Skinner, 1988). Thus, it is common to find that functional gain also reflects gain for conversational speech. On the other hand, in order for a nonlinear hearing aid (eg, a WDRC type with a low compression threshold) to provide the same wearer gain (insertion gain) at a conversational input level as a linear hearing aid, it would have to provide more gain for the lower input sounds.

Figure 13 shows a comparison of the static I/O curves between a linear and a nonlinear hearing aid when both are matched in output to a medium input level. It is easily seen that the nonlinear hearing aid yields more gain than the linear hearing aid for input levels below the medium input level. Thus, a lower aided threshold and a higher functional gain can be expected of the
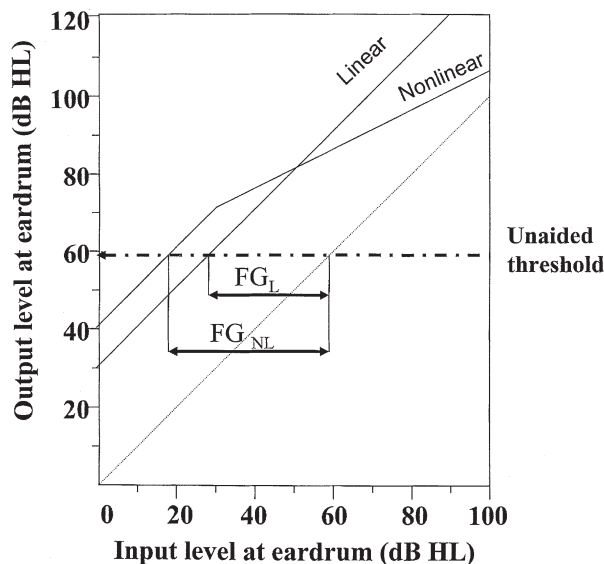
Figure 13. Hypothetical input/output curves of a linear and a nonlinear hearing aid matched in output to a medium (50 dB HL) input. Note that the nonlinear hearing aid has a higher output below the 50 dB HL input. This would result in a lower aided threshold and a higher functional gain.

nonlinear hearing aid than the linear hearing aid. In this case, the nonlinear hearing aid allows the wearer to hear sounds as soft as 20 dB HL, whereas the linear hearing aid allows the wearer to hear sounds at or greater than 30 dB HL when both have the same gain (30 dB) and output to an input level at 60 dB HL. Clinically, if a satisfied wearer of linear hearing aids is fitted with nonlinear hearing aids, one should expect to achieve a higher functional gain (or a lower aided threshold) with the nonlinear hearing aids than with the linear hearing aids if both hearing aids sound equally loud at a conversational input level. That is, they have the same insertion gain for normal sounds. Conversely, if both hearing aids yield the same aided threshold, it is likely that the nonlinear hearing aids may not be providing similar gain as the linear hearing aids for medium input level sounds (the extent is dependent on the compression ratio). In this case, the wearer complains that the nonlinear hearing aids are not loud enough even though the audibility for soft sounds is similar for both hearing aids.

Insertion gain reflects the gain (and output) of the hearing aid at a specific input level. It is not tied to the wearers' threshold perception and subjective participation is not necessary for its determination. However, variability and measurement errors would still be present (Dirks and Kincaid, 1987; Tecca *et al.*, 1987). An advantage of real-ear probe microphone measurement is that the output at various input levels can be determined. This is especially meaningful for the evaluation of nonlinear hearing aids where gain changes at different input levels can be studied or when the hearing aid output to a high input (eg, OSPL90) is examined.

In addition, REM can be used to examine special processing features on the hearing aids such as noise reduction, feedback cancellation, and directional microphone effects. Such information cannot be determined from functional gain or aided threshold measure. Functional gain reflects gain that the wearers receive; it does not, as was believed, reflect gain for conversational speech but rather reflects gain for a threshold response. Indirectly, it reflects the level of the softest sound that the wearers hear. On the other hand, insertion gain information does not reflect subjective perception or audibility. It is possible to determine insertion gain on the Knowles Electronics Manikin for Acoustic Research (KEMAR), but functional gain or the aided threshold cannot be measured on the same manikin. As the term implies, functional gain implies a functional improvement in auditory sensitivity whereas insertion gain implies a physical increase in gain. Although aided threshold or functional gain reflects only gain for the softest sounds, it is an important measure because without audibility, there can be no intelligibility.

REM and aided thresholds serve a complementary rather than an exclusive purpose. Thus, the two gain terms should not be derived from each other and they should not be used interchangeably or preferentially. Their relation can be analogous to the complementary use of forks and knives during meals. Forks are meant to pick up food, while knives are meant to cut food. One would not forgo the use of knives because they do not allow one to pick up food. Rather, one would use both in the normal course of a meal. The dining experience may be compromised if one is not used. The same is true for aided threshold and REM; both should be used in the verification of hearing aid performance, but for different purposes.

From a measurement standpoint, functional gain (and aided thresholds) can best be determined by the use of warble tones or narrow bands of noise in order to provide frequency-specific gain information. Use of broadband stimuli, such as speech-shaped noise, would not yield frequency-specific threshold information. This would be true for both linear and nonlinear hearing aids. On the other hand, any type of acoustic stimuli can be used to measure insertion gain, including sweep tones, speech-shaped noise, and white noise, to yield the same results if the hearing aids are linear. However, the same will not be true for nonlinear hearing aids, especially those with multiple channels. Different insertion gain may be reported depending on the nature of the stimulus and the processing of the nonlinear hearing aids, such as number of channels, compression ratios, and compression thresholds in the individual channels. The choice of stimulus for insertion gain determination in nonlinear hearing aids must be done judiciously (eg, Kuk and Ludvigsen, 2003).

Conclusions

The measurement of the aided sound-field threshold allows clinicians to estimate the lowest input level that is audible to the hearing aid wearer. Consequently, an aided threshold that is close to 20 dB HL should ensure audibility of soft speech for most fittings. This information makes the aided threshold a useful index to reflect the goodness of the fit of a nonlinear hearing aid for low input level sounds. Because this index is tied to the gain setting on a hearing aid, it is more meaningful as it applies to nonlinear hearing aids (than linear hearing aids) where the use of a volume control is minimized or not permitted. Because of potential measurement errors and efficiency, one may opt to predict the aided threshold if an in situ unaided threshold can be reliably estimated. Despite its usefulness, information provided by sound-field measures is not the same as that provided by probe microphone measures, especially in nonlinear hearing aids. Thus, these two indices must be determined separately for a complete verification of the wearer's performance with the hearing aids.

Appendix: Transformation from dB SPL to dB HL

The transformation from dB SPL to dB HL used in the illustration involves two steps. First, input/output data from its typical reference are transformed to a reference at the eardrum, ie, dB SPL(ED). Secondly, the data are converted to dB HL by including the SPL to HL conversion.

(a) Converting to Eardrum dB SPL, SPL (ED)

Imagine that a manikin such as Knowles Electronics Manikin for Acoustic Research (KEMAR) is placed in a free-field (FF) directly facing (ie, at 0°) the sound source that generates a sound pressure level of SPL(FF) when measured without the manikin. With the manikin but without a hearing aid on the manikin, the sound pressure that can be measured at the eardrum (ED) of KEMAR will be enhanced by the free-field-to-eardrum (FF2ED) transform. That is,

$$SPL(ED)_{unaided} = SPL (FF) + FF2ED \quad (A1)$$

If a hypothetical "transparent" hearing aid is placed on KEMAR, the sound pressure level at its eardrum will remain unchanged as in the unaided condition (ie, $SPL[ED]_{unaided}$) because the insertion gain of the transparent hearing aid by definition is 0 dB at all frequencies. The SPL at the inlet of the microphone of the transparent hearing aid is,

$$SPL(Mic) = SPL(FF) + FF2Mic \quad (A2)$$

Substituting Equation A2 into Equation A1, the SPL at the eardrum can be related to the sound pressure level at the microphone by Equation A3:

$$SPL(ED)_{THA} = SPL(Mic)$$
$$+ FF2ED - FF2Mic \quad (A3)$$

The appropriate FF2ED and FF2Mic (or microphone location effect, MLE) corrections can be determined from Bentler and Pavlovic (1989). The combined transfer function relating SPL at the microphone of a transparent behind-the-ear (BTE) hearing aid to SPL at the eardrum is shown in Figure A1.

To convert the output from a 2-cc coupler reference to the eardrum reference, the 2-cc-to-eardrum (2cc2ED) correction (Sachs and
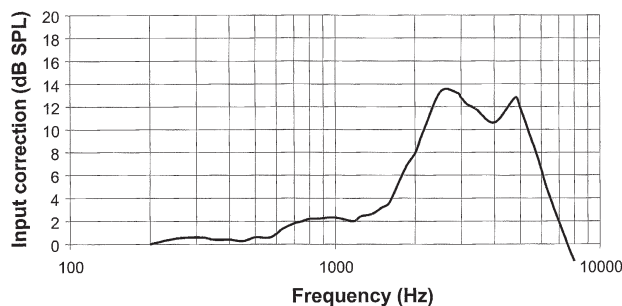
Figure A1. Transfer function for transforming sound pressure level at a behind-the-ear microphone input to sound pressure level measured at the eardrum for an average wearer (combination of free-field-to-eardrum and free-field-to-microphone corrections as reported in Bentler and Pavlovic, 1989). Abbreviations: SPL, sound pressure level.
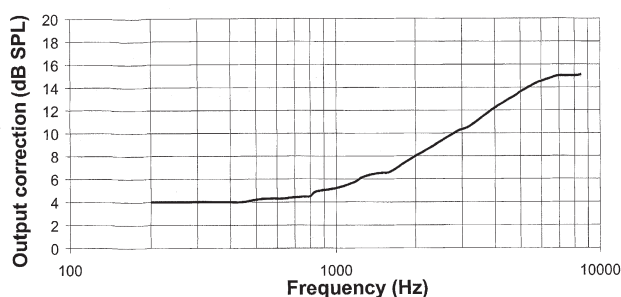
Figure A2. Transfer function for transforming sound pressure level measured in a 2-cc coupler to sound pressure level measured at the ear drum (2cc2ED). Data from Bentler and Pavlovic (1989). Abbreviation: SPL, sound pressure level.

Burkhard, 1972) shown in Figure A2 will have to be added.

$$SPL(ED) = SPL (2cc) + 2cc2ED \qquad (A4)$$

(b) Converting to dB HL

To further simplify the representation , the input and output can be expressed in dB HL instead of dB SPL. The threshold of audibility (0 dB HL) corresponds to the dB SPL required to reach threshold in an average adult ear (Sivian and White, 1933). Thus, to convert from dB SPL to dB HL,

one simply subtracts from the dB SPL either the minimum audible pressure (MAP) when reference is made to the eardrum or the minimum audible field (MAF) when reference is made to a free field. Figure A3 shows the MAP and MAF data from Bentler and Pavlovic (1989).

Because in hearing aid literature, it is customary to reference SPL to eardrum position, the input to a transparent hearing aid expressed in dB HL is:

$$HL = SPL(Mic) + FF2ED$$
$$- FF2Mic - MAP \qquad (A5)$$

Alternatively, the same exact outcome will result if one transforms SPL using the free-field reference by subtracting MAF. In this case, if MAF is subtracted from both sides of Equation A2, the result is:

$$HL = SPL(FF) - MAF = SPL(Mic)$$
$$- FF2Mic - MAF \qquad (A6)$$

Because MAF = MAP-FF2ED, replacing MAF in Equation A6 by MAP-FF2ED will yield the same result as shown in Equation A5. Thus the dB HL notation is independent of the reference position, unlike the dB SPL notation.

Whereas the output expressed in dB HL for any hearing aid is:

$$HL = SPL(2cc) + 2cc2ED - MAP \qquad (A7)$$
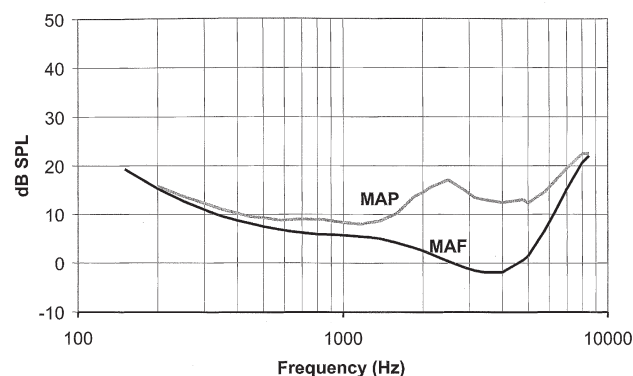
Figure A3. Threshold estimates obtained by minimum audible pressure (MAP) and minimum audible field (MAF). Data from Bentler and Pavlovic (1989). Abbreviation: SPL, sound pressure level.

References

Output (at 1 K Hz)
SPL (ED) = SPL (2cc) +2cc2ED
         = SPL (2cc) + 5.2
HL (ED) = SPL (ED) - MAP
        = SPL (ED) - 8.3
HL (ED) = SPL (2cc) –3.1

Input (at 1 KHz)
SPL (ED) = SPL (mic) - FF2Mic +FF2ED
         = SPL (mic) -0.3 +2.6
HL (ED) = SPL (ED) - MAP
        = SPL (ED) - 8.3
HL (ED) = SPL (Mic) - 6
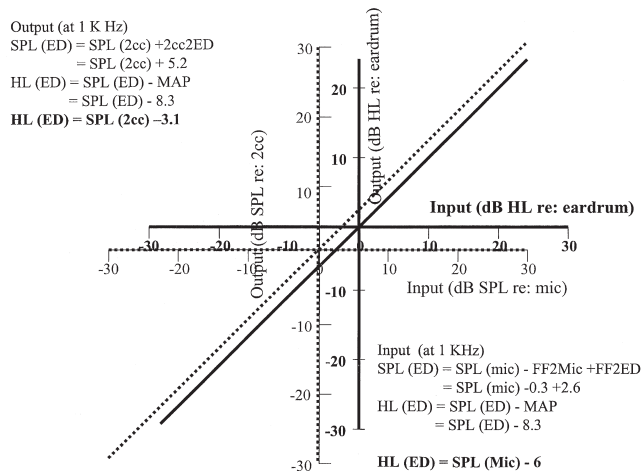
Figure A4. Input/output graphs using different reference sources. The thin dotted axes reference the sound pressure level (SPL) measured at the microphone as the input and at the 2-cc coupler as the output (in dB SPL). The thick solid axes reference the sound pressure level measured at the eardrum as the input and as the output (both in dB HL). Note the consequence of the conversion (from dB SPL to dB HL) is a shift in the position of the X- and Y-axes. Abbreviations: FF2Mic, free-field-to-microphone; FF2ED, free-field-to-eardrum.

As an illustration, Figure A1 shows that the combined correction of (FF2Mic - FF2ED) at 1000 Hz is 2.3 dB, and Figure A3 shows that MAP at 1000 Hz is 8.3 dB. Equation A5 would suggest that a microphone input of 6 dB SPL corresponds to an input of 0 dB HL. By the same token, Equation A7 would suggest that a 2-cc coupler output of 3.1 dB SPL is needed in order to reach an eardrum output of 0 dB HL at 1000 Hz when the 2cc2ED correction is 5.2 dB (shown in Figure A2). Based on Equations A5 and A7, the relationship between the two references (dB SPL at microphone in dotted line, and dB HL in dark line) can be illustrated with the line of unity gain (no gain) drawn through their respective origins (Figure A4). As another example, an input of 10 dB HL and an output of 10 dB HL would correspond to an input of 16 dB SPL measured at the microphone opening and an output of 13.1 dB SPL measured in a 2-cc coupler when the hearing aid provides unity (or no) gain. It is seen that the I/O curve, when expressed in dB HL, is simply a shift of the scales without any alteration of the I/O function.

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